The power of standing up

Regterschot, Gerrit

CHAPTER 5

Accuracy and concurrent validity of a sensor-based analysis of sit-to-stand movements in older adults

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ABSTRACT

Body-fixed motion sensors have been applied for the assessment of sit-to-stand (STS) performance. However, validity of some sensor-based STS measures is unclear. Therefore, this study investigated accuracy and concurrent validity of sensor-based STS measures in older adults. Twenty-seven older adults (20 females, 7 males; age: 72-94 years) performed five STS movements while data were collected with force plates and motion sensors on the hip and chest. Hip maximal acceleration provided an accurate estimation of the center of mass (CoM) maximal acceleration (limits of agreement (LOA) smaller than 5% of the CoM maximal acceleration; estimated and real CoM maximal acceleration did not differ ($p=0.823$)). Other hip STS measures and the chest STS measures did not provide accurate estimations of CoM motion (LOA ranged from -155.6% to 333.3% of the CoM value; sensor-based measures overestimated CoM motion (range p: <0.001 to 0.01)). However, the hip sensor did not overestimate maximal jerk of the CoM ($p=0.679$). Moderate to very strong associations were observed between sensor-based estimations and actual CoM motion (range $r=0.64$-0.94, $p<0.001$). Hence, sensor-based estimations of CoM motion during STS are possible, but accuracy is limited. The sensor-based method may be relevant for clinical assessments, however, it cannot replace laboratory methods for a biomechanical analysis of STS.

1. INTRODUCTION

Leg muscle power is a determinant of movement execution and an important parameter for measuring intervention effects in older adults [1-4]. However, available methods for the measurement of leg muscle power, such as force plates, cycle ergometers and isokinetic dynameters, have practical disadvantages that limit the application of these methods in clinical settings. For example, the aforementioned methods for the measurement of leg muscle power are expensive, difficult to transport and not easy to use. Therefore, it is important that new, practical methods are developed for the measurement of leg muscle power in older adults.

Zijlstra et al. (2010) developed an alternative method for the measurement of leg muscle power based on small body-fixed motion sensors [5]. In this study body-fixed motion sensors were used to estimate the vertical peak power of the body's center of mass (CoM) during the sit-to-stand (STS) transfer. Results indicated fair to excellent concurrent validity of sensor-based estimations of CoM peak power during STS based on a comparison with force plate measurements. In addition, results showed that a sensor on the hip provided more accurate estimations of vertical CoM acceleration during STS than a sensor on the chest, which overestimated CoM accelerations and peak powers [5].

Recent studies developed sensor-based measures of STS in addition to STS peak power (e.g. maximal vertical velocity, maximal vertical acceleration during STS) [6,7]. However, the accuracy and concurrent validity of these additional sensor-based STS measures are unclear. For the interpretation of such measures, it is important that accuracy and concurrent validity of sensor-based STS measures are investigated in older adults. When concurrent validity is adequate, sensor-based estimations of real STS performance are possible. Furthermore, when accuracy is adequate, the sensor-based method may replace standard laboratory methods for the biomechanical analysis of STS. Therefore, the aim of this study was to investigate the accuracy and concurrent validity of sensor-based STS measures in older adults. For this purpose, we compared the sensor-based method to a standard laboratory method consisting of force plates under the chair and feet of the participants. Based on the findings of Zijlstra et al. (2010) [5], we hypothesized that hip STS measures have adequate concurrent validity and accuracy, and that chest STS measures have adequate concurrent validity but overestimate CoM kinematics resulting in inadequate accuracy.

2. METHODS

2.1 Participants

Participants were recruited from a health care center, a residential care home and sheltered houses. Older adults could participate in this study when they were able to rise from a chair, walk at least 10 m (with or without a cane or wheeled walker), and when they were at least 70 years of age. Participants were excluded when they had any cognitive, neurological, cardiovascular or respiratory disorder, lower extremity orthopaedic surgery or a stroke within the six months before the study, severe comorbidity, significantly reduced vision.

In this study 27 older adults (20 females) participated on a voluntary basis. Age ranged from 72-94 years (81.7±5.6 years), body mass ranged from 48.0-98.9 kg (75.7±13.3 kg), and body height was between 1.46-1.84 m (1.63±0.09 m).

The present study was approved by the Medical Ethical Committee of the University Medical Center Groningen, the Netherlands (METc2011.054). The study protocol is in agreement with the Helsinki declaration. An informed consent was signed by all participants.

2.2 Procedures

Participants performed five chair rise movements at a normal speed from a standard chair (height: 0.47m). Prior to standing up participants were leaning against the back of the chair. Participants stood up with their arms crossed in front of the chest. After rising from the chair participants stood still for 5 seconds before sitting down again. Between each stand-to-sit and sit-to-stand movement participants sat still on the chair for 10 seconds.

2.3 Data acquisition

2.3.1 Body-fixed motion sensors

During the sit-to-stand movements participants wore two body-fixed motion sensors (π-Node, Philips). Both sensors consisted of a 3D accelerometer (±2g), a 3D gyroscope (±300°/s) and a 3D magnetometer (±2G) [8]. One sensor was worn on the right side of the hip (just above the trochanter major femoris; see Figure 2 in Regterschot et al., 2014 [6]) because a previous study demonstrated that a sensor at this location provides a more accurate estimation of the vertical CoM acceleration during STS than sensors at other locations [5]. The other sensor was worn on the chest (sternum) because this location seems preferable for activity monitoring [9]. Hereafter we refer to the sensors as hip sensor and chest sensor. Data were collected with 50 Hz sampling frequency and wirelessly transmitted to a PC for storage [8].

2.3.2 Force plates

Measurements were performed with two force plates (Bertec; each plate measured 0.60 m x 0.40 m). One force plate was positioned under both feet of a participant, the other force plate was located under the chair. Force plate data were sampled with 100 Hz frequency.

2.4 Data processing

Processing of the sensor data and the force plate data was performed using Matlab (The Mathworks, Inc.; version 7.12).

2.4.1 Body-fixed motion sensors
Quaternions were applied to estimate the accelerations of the hip sensor and the chest sensor in the global coordinate system using the accelerometer, gyroscope and magnetometer data in the sensor coordinate system [8]. Data were filtered with a low-pass Butterworth filter (cut-off frequency of 3 Hz [5]).

2.4.2 Force plates
The vertical data of both force plates were filtered with a low-pass Butterworth filter (cut-off frequency of 3 Hz [5]). Subsequently the vertical force data of both force plates were summed to calculate the vertical force of the body's CoM (F_{com}). Vertical acceleration of the body's CoM (a_{com}) was computed by applying the following formula: $a_{com} = F_{com} / m$. In this formula m represents body mass.

2.5 Data analysis

Data analysis was performed using Matlab (The Mathworks, Inc.; version 7.12). The vertical acceleration data of the hip sensor in the global coordinate system (a_{hip}), the vertical acceleration data of the chest sensor in the global coordinate system (a_{chest}), and the vertical acceleration data of the body's center of mass as determined based on force plates (a_{com}) were separately used for the calculation of the following STS measures:

1. STS duration (s): Interval between the initiation of the forward trunk rotation prior to STS and the first intersection of the vertical acceleration data with the gravitational acceleration, after the deceleration phase (see Figure 3 in Regterschot et al., 2014 [6]).
2. Maximal acceleration (m/s^2): Maximal vertical acceleration during STS.
3. Maximal jerk (m/s^3): Maximal positive jerk during the acceleration phase of the STS movement. Jerk was calculated as: $jerk_i=(a_{i+1}-a_i)/(1/fs)$ with i indicating sample number, a vertical acceleration, and fs sampling frequency.
4. Maximal velocity (m/s): Maximal vertical velocity during STS. Velocity was estimated by numerical integration of the vertical acceleration during STS. We assumed that vertical velocity was 0 m/s at the initiation of STS.
5. Peak power (W): Maximal vertical power generated during STS. Force (F) and velocity (v) were multiplied to estimate power: $P_i=F_i \cdot v_i$ [5]. Force (F) was computed using: $F_i=m \cdot a_i$. In this formula m represents body mass and i indicates sample number.
6. Scaled peak power (dimensionless): Peak power corrected for body mass (m), body

height (l) and gravity (g): $P_{scaled}=P/(m \cdot g^{1.5} \cdot l^{0.5})$ [10].
7. SD stabilization phase (m/s²): SD of the vertical acceleration data during the stabilization phase. Since in most older persons the duration of the stabilization phase is shorter or equal to 0.8s [11], we defined the stabilization phase as the interval of 0.8s following STS.

2.6 Statistical analysis

The mean value of the STS variables was calculated across the five STS trials, since a previous study showed that the mean value across five STS trials resulted in the highest test-retest reliability for most sensor-based STS variables [7].

Accuracy was examined using Bland and Altman agreement tests [12]. Mean differences between the sensor method and the force plate method were calculated as well as SD of differences and limits of agreement (LOA). The LOA were calculated as: LOA = D ± 1.96 * SD_{diff}. In this formula D represents the mean difference between the sensor method and the force plate method, and SD_{diff} represents the SD of the differences between the sensor method and the force plate method. The LOA indicate a range within which two methods can be used interchangeably [12]. Sensor-based STS measures were considered accurate when both the upper and lower LOA were smaller than 10% of the average of the real CoM values as measured with the force plates, and when there were no systematic differences between sensor-based estimations and real CoM values. Systematic differences between the sensor method and the force plate method were investigated using dependent t-tests.

Concurrent validity of sensor-based STS measures was investigated using Pearson correlation coefficients. Pearson correlation coefficients (r) were calculated between sensor-based STS measures and STS measures calculated from force plate data. Correlations were interpreted as follows: little (if any correlation) when $0.00<r\leq0.25$; weak when $0.26\leq r\leq0.49$; moderate when $0.50\leq r\leq0.69$; strong when $0.70\leq r\leq0.89$; very strong when $0.90\leq r\leq1.00$ [13]. Statistical significance was set at $p<0.05$. Statistical analyses were performed with SPSS Statistics (IBM; version 20).

3. RESULTS

Missing samples were observed in the chest sensor data of three participants. Therefore these data were excluded from further analysis.

3.1 Accuracy

Table 1 shows the outcomes of the Bland and Altman agreement tests and the dependent t-tests. In Figure 1 Bland and Altman plots are visible for each STS variable. Only the LOA of hip maximal acceleration were smaller than 10% of the CoM values (Table 1). LOA of the other hip STS measures ranged from -74.3% to 125.0% of the CoM values (Table 1). LOA of the chest STS measures ranged from -155.6% to 333.3% of the CoM values (Table 1). For all chest STS measures, except STS duration, Bland and Altman plots show a positive linear relationship (range R^2: 0.65-0.83), but not for the hip STS measures (Figure 1).

Systematic differences between the sensor method and the force plate method were observed for all STS measures (range p-values: <0.001 to 0.01), except for hip maximal acceleration (p=0.823) and hip maximal jerk (p=0.679) (Table 1). The systematic differences consisted of an overestimation by the sensor methods compared to the force plate method (Table 1).

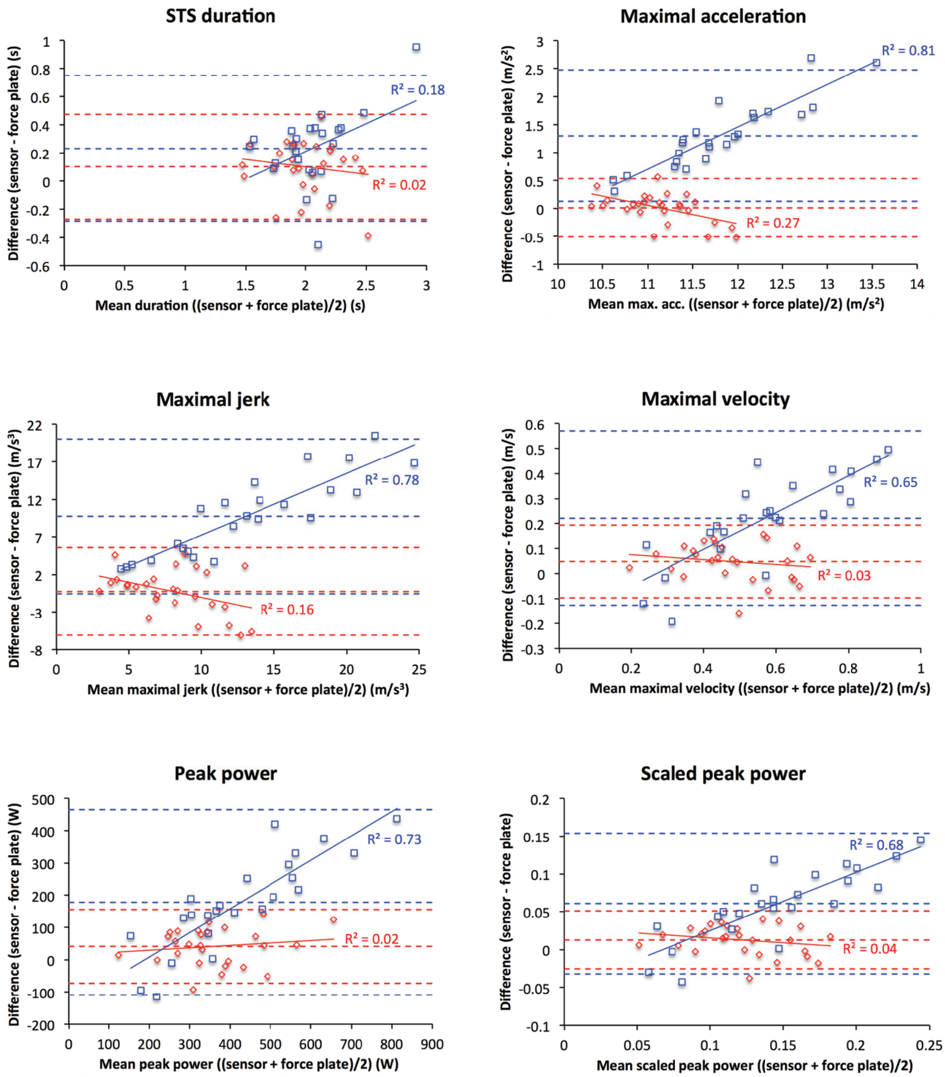
3.2 Concurrent validity

Table 2 shows the associations between the sensor method and the force plate method. Hip SD stabilization phase, chest SD stabilization phase and chest maximal acceleration showed a very strong association with similar force plate measures (range r=0.92-0.94; Table 2). Hip STS duration, hip maximal acceleration, hip maximal velocity, hip peak power, hip scaled peak power, chest maximal jerk, chest maximal velocity, chest peak power and chest scaled peak power demonstrated a strong association with similar force plate measures (range r=0.79-0.88; Table 2). Hip maximal jerk and chest STS duration showed a moderate association with similar force plate measures (range r=0.64-0.69; Table 2).

Table 1 | Mean±SD of sensor measurements and force plate measurements. Outcomes of agreement tests and dependent t-tests are also shown.

	Sensor	Force plate	$Mean_{diff}$	SD_{diff}	Lower LOA (%[a])	Upper LOA (%[a])	t	p-value[b]
Hip sensor (n=27)								
STS duration (s)	2.05±0.27	1.95±0.30	0.10	0.19	-0.27 (-13.8%)	0.47 (24.1%)	-2.776	0.010*
Maximal acceleration (m/s^2)	11.15±0.37	11.13±0.51	0.01	0.27	-0.51 (-4.6%)[c]	0.53 (4.8%)[c]	-0.226	0.823
Maximal jerk (m/s^3)	7.94±2.77	8.18±3.87	-0.24	2.98	-6.08 (-74.3%)	5.60 (68.5%)	0.419	0.679
Maximal velocity (m/s)	0.50±0.13	0.45±0.14	0.05	0.07	-0.10 (-22.2%)	0.19 (42.2%)	-3.348	0.002*
Peak power (W)	379.8±121.0	338.5±112.8	41.3	58.2	-72.7 (-21.5%)	155.3 (45.9%)	-3.693	0.001*
Scaled peak power	0.13±0.03	0.12±0.04	0.01	0.02	-0.03 (-25.0%)	0.05 (41.7%)	-3.379	0.002*
SD stabilization phase (m/s^2)	0.11±0.10	0.08±0.09	0.03	0.04	-0.05 (-62.5%)	0.10 (125.0%)	-3.196	0.004*
Chest sensor (n=24)								
STS duration (s)	2.17±0.37	1.94±0.26	0.23	0.26	-0.29 (-14.9%)	0.75 (38.7%)	-4.306	<0.001*
Maximal acceleration (m/s^2)	12.44±0.99	11.14±0.46	1.30	0.60	0.13 (1.2%)	2.48 (22.3%)	-10.634	<0.001*
Maximal jerk (m/s^3)	17.91±8.08	8.17±3.56	9.74	5.25	-0.54 (-6.6%)	20.03 (245.2%)	-9.094	<0.001*
Maximal velocity (m/s)	0.68±0.27	0.46±0.13	0.22	0.18	-0.13 (-28.3%)	0.57 (123.9%)	-6.078	<0.001*
Peak power (W)	514.9±232.5	336.8±110.7	178.1	146.4	-108.7 (-32.3%)	465.0 (138.1%)	-5.962	<0.001*
Scaled peak power	0.18±0.07	0.12±0.03	0.06	0.05	-0.03 (-25.0%)	0.15 (125.0%)	-6.265	<0.001*
SD stabilization phase (m/s^2)	0.17±0.20	0.09±0.10	0.08	0.11	-0.14 (-155.6%)	0.30 (333.3%)	-3.657	0.001*

[*] $p<0.05$ indicates statistical significance.
[a] LOA expressed as a percentage of the mean of real CoM values as measured with force plates.
[b] Two-tailed p-value.
[c] Upper and lower LOA are smaller than 10% of the mean of real CoM values measured with force plates.



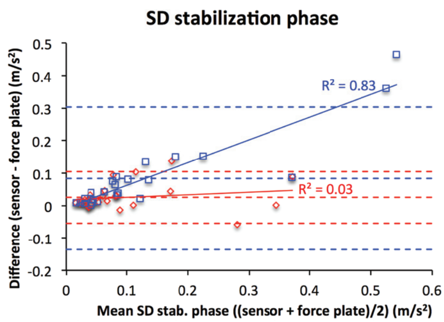


Figure 1 | Bland and Altman plots showing the differences between the sensor method and the force plate method for each STS variable. Hip sensor vs force plate is shown in red color, chest sensor vs force plate is shown in blue color. Dashed lines indicate the mean difference and the LOA. Also a linear trend line is shown with the coefficient of determination (R^2).

Table 2 | Pearson correlation coefficients (r) between outcomes of sensor measurements and outcomes of force plate measurements.

	Hip sensor (n=27)		**Chest sensor (n=24)**	
	r	**p-value[a]**	**r**	**p-value[a]**
STS duration (s)	0.79	<0.001*	0.69	<0.001*
Maximal acceleration (m/s^2)	0.86	<0.001*	0.92	<0.001*
Maximal jerk (m/s^3)	0.64	<0.001*	0.88	<0.001*
Maximal velocity (m/s)	0.86	<0.001*	0.83	<0.001*
Peak power (W)	0.88	<0.001*	0.87	<0.001*
Scaled peak power	0.85	<0.001*	0.83	<0.001*
SD stabilization phase (m/s^2)	0.92	<0.001*	0.94	<0.001*

[*] p<0.05 indicates statistical significance.
[a] Two-tailed p-value.

4. DISCUSSION

This study investigated the accuracy and concurrent validity of sensor-based STS measures in older adults by comparing the sensor-based method to a standard laboratory method consisting of force plates. Results showed that only hip maximal acceleration has adequate accuracy in older adults. The other hip STS measures and chest STS measures have inadequate accuracy and overestimated CoM motion. Hip maximal jerk showed inadequate accuracy, however, this measure did not show a systematic difference with the maximal jerk of the CoM. Adequate concurrent validity was observed for all sensor-based STS measures. In particular hip SD stabilization phase, chest maximal acceleration and chest SD stabilization phase showed very strong associations with CoM values (range r=0.92-0.94). The other sensor-based STS measures demonstrated associations with CoM values that ranged from moderate to strong (range r=0.64-0.88). Together these results almost completely confirm our hypothesis, however, results do not support adequate accuracy of all hip STS measures.

The overestimation of CoM peak power by the chest sensor was also observed by Zijlstra et al. (2010) [5]. However, the overestimation of CoM peak power by the hip sensor was not observed by Zijlstra et al. (2010) [5]. Zijlstra et al. (2010) explained the overestimation of CoM peak power by the chest sensor based on trunk kinematics [5]. The upper trunk shows an upward movement during STS which consists of the upward movement of the CoM and the hip extension movement. This results in higher vertical accelerations at the chest compared to the vertical accelerations of the CoM. Therefore, a sensor at the chest overestimates vertical CoM accelerations. This explains the overestimation of maximal acceleration, maximal jerk, maximal velocity, peak power and scaled peak power by the chest sensor in the present study. Furthermore, it seems very likely that the better the STS performance, the higher the accelerations as measured by the chest sensor deviate from CoM accelerations as measured by force plates. The Bland and Altman plots support this assumption, since maximal acceleration, maximal jerk, maximal velocity, peak power and scaled peak power measured with the chest sensor show larger differences with force plate outcomes in individuals with a better STS performance.

An additional explanation for the overestimation of CoM peak power, scaled peak power and maximal velocity by the chest sensor may be the use of a different starting point of the integration interval compared to the force plate method. A different starting point of the integration interval may also explain the overestimation of CoM peak power, scaled peak power and maximal velocity by the hip sensor. In addition, a difference in the detection of the starting point of STS may explain the overestimation of STS duration by the sensor methods compared to the force plate method. However, the overestimation of STS duration by the sensor methods may also be the result of a difference in the ending point of STS between the sensor methods and the force plate method. Moreover, a difference in the detection of the ending point of STS may explain the overestimation of SD stabilization phase by the sensor methods compared to the force plate method, because the interval of 0.8s after the ending point of STS was used for the calculation of SD stabilization phase. Since we did not synchronize the sensors and the force plates, it is not possible to determine whether both the starting point and ending point differed between the sensor methods and the force plate method. For this reason, future research should synchronize sensors with force plates.

The associations between sensor-based STS peak power and CoM peak power calculated based on force plates were comparable to the associations found by Zijlstra et al. (2010) [5]. In the present study the associations with CoM peak power were r=0.88 for hip peak power and r=0.87 for chest peak power. In the study of Zijlstra et al. (2010) the associations with CoM peak power were respectively r=0.94 for hip peak power and r=0.93 for chest peak power [5]. In the present study there were only small differences in associations with force plate outcomes between the two sensors. Largest differences were observed between hip STS duration (r=0.79) and chest STS duration (r=0.69), and between hip maximal jerk (0.64) and chest maximal jerk (r=0.88).

The overestimation of actual CoM motion by most sensor-based STS measures has consequences for the interpretation of sensor-based STS measures. Actual CoM motion during STS is smaller than the outcomes of most sensor-based STS measures indicate. This is particularly true for the chest sensor of which most STS measures show increasing overestimations with better STS performance. Since the sensor methods have limited accuracy for the estimation of CoM motion during STS, the body-fixed motion sensors cannot replace laboratory methods for a biomechanical analysis of STS. An adequate biomechanical analysis of CoM motion during STS requires laboratory methods, such as force plates or camera systems. However, the sensor-based method may be relevant for clinical assessment of STS performance, because the present study shows that sensor-based STS measures are associated with CoM motion during STS and other studies revealed that specific sensor-based STS measures have adequate test-retest reliability, sensitivity to change and discriminative ability in older adults [6,7,14,15].

The STS performance of the older adults as measured with the force plates was comparable to the STS performance of older adults measured with force plates in other studies. The range of CoM peak powers in the present study (117.0-594.3 W) includes the mean CoM peak power as reported in other studies (439.5±158.8 W [5], 457.37±142.69 W [11] and 424±161 W [16]). In addition, the range of STS durations measured with force plates in the present study (1.41-2.71 s) includes the mean STS duration as measured with force plates in another study (1.82±0.51 s [11]). Furthermore, the range of maximal velocities measured with force plates in the present study (0.18-0.69 m/s) includes the mean maximal velocity measured with force plates in another study (0.45±0.13 m/s in an exercise group and 0.49±0.24 m/s in a control group during pre-measurement [17]).

However, the present study has several limitations. First, the motion sensors and force plates were not synchronized. As a consequence, it remains unclear whether both the starting point and ending point of STS differed between the sensors and the force plates. Another limitation of the study is that we did not use camera data. By using position data we could have analyzed the trunk movement more precisely. This would have helped interpreting the results but it would particularly have enabled us to determine power for trunk vertical displacement which is not equal to CoM power. Such a comparison might have shown a better correspondence than the present comparison between the sensor method and the force plate method. Another limitation of this study was the data loss from the chest sensor during STS assessments in three participants, however, it is unlikely that this limitation has a significant effect on the findings of the present study. In future studies this problem can be circumvented by using a motion sensor with a possibility for local data storage (see e.g. [14]). An additional limitation of this study was the fact that we evaluated sensor-based estimations from only two body locations. More accurate estimations of CoM motion may be possible at other body locations or by using a weighted average of the hip and chest sensor [5]. However, for practical reasons a single-sensor approach is preferred over a multi-sensor approach.

5. CONCLUSIONS

In conclusion, this study demonstrated that only maximal acceleration of the CoM can be estimated with adequate accuracy and adequate concurrent validity by a body-fixed sensor on the hip. Other aspects of CoM motion during STS can be estimated with adequate concurrent validity by body-fixed sensors, but accuracy is limited. Hence, the sensor-based method may be a practical alternative for the clinical assessment of STS performance in older adults, however, the sensor-based method cannot replace standard laboratory methods for a biomechanical analysis of CoM motion during STS.

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